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(54) IMPROVEMENTS IN OR RELATING TO RESPIRATORS

(71) We, F. HOFFMAN-LA ROCHE & CO. AKTIENGESELLSCHAFT, a Swiss company of 124-184 Grenzacherstrasse, Basle, Switzerland, do hereby declare the invention, for which we pray that a patent may be granted to us, and the method by which it is to be performed, to be particularly described in and by the following statement:—

This invention relates to a respirator, more particularly a respirator in which the flow and pressure of the respiratory gas is automatically controlled during inhalation and exhalation.

Respirators of the kind described hereinabove are used to replace or assist the respiratory function in patients in which spontaneous respiration is either absent or insufficient. Respirators of this kind may also be employed for anaesthesia.

Conventional respirators represent controlled systems, that is to say their function takes place in accordance with predefined input variables. One group of known respirators comprises the so-called "pressure-controlled system" by means of which respiratory gas under pressure is supplied to the patient, a control valve interrupting the supply of respiratory gas when a specific predefined pressure has been built up in the duct which extends to the patient. A further group comprises the so-called "volumetrically controlled system" by means of which a volumetrically measured quantity of the respiratory gas is supplied to the patient. Both kinds of respirators suffer from a disadvantage which is basically common to all controlled systems: they are unable to react in a compensating manner to changes of the mechanical lung action which cannot be predetected.

Automatically controlled respirators have also been proposed but for various reasons these are not suitable for general, routine application in hospitals. One of the reasons is the fact that measurement of the controlled condition calls for apparatus which is complicated and is, therefore, on the one hand expensive and on the other hand trouble-prone.

It is the object of the present invention to provide an automatically controlled respirator which overcomes or at least mitigates the disadvantages of known systems and which, because of its generally simple and inexpensive production is generally suitable for a wide range of applications.

According to the present invention there is provided a respirator with automatic control for flow and pressure of respiratory gas, which respirator comprises a flow and pressure measuring device disposed directly adjacent to a patient connection, the device being for measuring the flow and pressure of respiratory gas and for converting these parameters into electrical signals, a valve system disposed between the respiratory gas source and the flow and pressure measuring device for controlling the flow and pressure of respiratory gas during inhalation and exhalation, and an electric control device comprising the measuring device, the valve system and the electric control device being interconnected to constitute a control loop for comparing the electrical signals generated in use of the respirator with set values and for generating a correcting signal for influencing the valve system.

Reference is directed to concurrently filed Application no. 14561/73 (Serial No. 1432234) which describes and claims a device having a valve function and by means of which a stream of gas or liquid may be conducted from a first connection to a second connection, whereas a stream orientated oppositely through the second connection is conducted to a third connection, which device comprises a first duct extend-

ing from the first connection to the second connection, a second duct disposed perpendicular to the first duct and adapted to connect the first duct to the third connection and a flow-immersed member disposed co-axially relative to the first duct and disposed opposite the opening of the second duct into the first duct, the flow immersed member having a middle cylindrical part, a frusto-conical part adjoining each end of the cylindrical part and a duct extending from the endface of the flow-immersed member facing the second connection to the cylindrical part of the external surface of the flow-immersed member disposed opposite to the position at which the second duct joins the first duct.

For a better understanding of the present invention and to show how the same may be carried into effect, reference will now be made, by way of example, to the accompanying drawings, in which:

FIGURE 1 shows a block diagram of an automatically controlled respirator according to the present invention,

FIGURE 2 shows a perspective view of a preferred embodiment of the regulating valve, of the inhalation/exhalation valve and of the measuring head in the assembled state, of the respirator of Figure 1,

FIGURE 3 shows a partial sectional view of the regulating valve, along planes A-A of Figure 2,

FIGURE 4 shows a longitudinal sectional view along the plane B-B of Figure 2,

FIGURE 5 shows a diagrammatic elevational view of the measuring head of Figure 2,

FIGURE 6 shows a sectional view of a detail of Figure 4 to an enlarged scale,

FIGURE 7 shows a block diagram of the electronic regulating device of the respirator of Figure 1,

FIGURE 8 shows a block diagram of the circuit for the automatic zero setting of the flow signal of the respirator of Figure 1, and

FIGURE 9 shows a graph of voltage curves at some points of the circuit illustrated in Figure 8.

Referring now to the drawings, the respirator shown in the block diagram of Figure 1 and partially in Figure 2 in perspective is designed to supply respiratory gas from a respiratory gas source (not shown) at a specific rhythm similar to natural respiration, to a patient 11 and to discharge such gas from the pateient (the patient 11 naturally not forming part of the apparatus *per se*). A pressure generator or a central supply system of the kind usually installed in large hospitals may be used as the respiratory gas source. Such a system may provide any kind of different gas or gas mixtures, for example pure respiratory gases such as air, oxygen or mixtures with anaesthetic gas such as ether, nitrous oxide or the like. The respiratory gas supplied in this way should preferably have a pressure of 1.2 atm abs. Furthermore, the moisture and temperature of the respiratory gas should be an optimum match to respiration requirements, that is to say it should have a temperature of 37°C and should be saturated with more than 90% water. The devices required to this end are well known in the art.

A duct 1 is provided to connect the respirator to the respiratory gas source and extends to a regulating valve 2. The regulating valve 2 is provided to change over between the inhalation phase and the exhalation phase and at the same time for controlling the flow and pressure in the two control positions. In the inhalation phase, the valve 2 connects the duct 1 to a duct 3 and in the exhalation phase it connects the duct 1 to a duct 4. The magnitude of pressure and flow in both phases is controlled by greater or lesser opening of the valve. The two ducts 3 and 4 extend to an inhalation/exhalation valve 5, referred to hereinafter briefly as I/E valve. A patient duct 6 and an exhalation duct 7 are connected to the I/E valve in addition to the two ducts 3 and 4. The duct 7 may either be open to atmosphere or may have a duct 8 which returns to the respiratory gas source for the purpose of recovering the exhalation gas.

The patient duct 6 extends through a measuring head 9, for measuring the flow and pressure of the respiratory gas, directly to tracheal tube for intubation of the patient. The measuring head 9 generates electrical signals which represent the flow and pressure changes of the respiratory gas. An electrical conductor 12 transmits the signals thus obtained to an electronic control system 13. The control system 13 is supplied via a conductor 14 with signals corresponding to the desired values for the respiratory action. The measured signals are processed in the control system 13, are compared with the desired value signals to provide a correcting signal for controlling the regulating valve 2 to effect that measured values follow the desired values. An electric output conductor 15 transmits the correcting signals to the drive mechanism of the regulating valve 2.

As may be seen particularly by reference to Fig. 2, the individual elements of the controlled system, that is to say the regulating valve, the I/E valve and the measuring head are combined directly into an integral unit. This unit is exceptionally compact its dimensions being approximately 15 × 10 × 5 cm.

The regulating valve 2, the I/E valve

5, the measuring head and the control system 13 will be described in detail hereinbelow.

The regulating valve 2 is shown partially in section in Fig. 3. A substantially hollow cylindrical two portion casing 16, 18 has an inner bulkhead 17 extending perpendicularly to its axis, one side of the bulkhead being provided with a commercial servomotor (not shown). The casing portion 18 within which the motor is disposed is appropriately provided with cooling fins 19. The end of casing portion 18 is closed by a lid (not shown). The motor spindle (not shown) is extended beyond the lid.

The second portion 16 of the casing contains the actual valve part. The motor spindle is connected to a rotor 21 of hollow cylindrical shape with stepped diameter and closed at its smaller end. The external diameter of the widest cylindrical part of the rotor is slightly smaller, for example by 0.1 mm, than the internal diameter of the casing. A circular bore 22 is provided at one position of the cylindrical wall of the rotor 21. The wall of the casing portion 16 has two ports 23 and 24. The axes of the bore 22 of the rotor and of the two ports 23 and 24 of the casing are in a plane which is perpendicular to the casing axis. The distance between the centres of the two ports 23 and 24 of the casing is greater than the diameter of the bore 22. Complete or partial coincidence between the rotor bore 22 with one of the two casing ports 23 or 24 is obtained by appropriate angular position of the rotor, or the casing ports 23 and 24 are closed by the rotor wall.

The portion 16 of the casing which contains the rotor, that is to say the actual valve part, is disposed in an appropriate bore of a plastics block 25 which contains the inlet duct 1, extending to the regulating valve 2, and the outlet ducts 3 and 4 which extend away from the regulating valve, the said block being also part of a housing containing the I/E valve in the measuring head as will be explained hereinbelow. The inlet duct 1 extends into the interior of the casing portion 16, and of the rotor while the outlet ducts 3 and 4 provide communication between the casing ports 23 and 24 and the appropriate inlets of the I/E valve.

The plastics material block 25 is detachably joined to the casing 16, that is to say for example with clips or a bayonet closure.

A port 26 is provided in the casing wall to vent the space between the rear of the rotor 21 and the bulkhead 17.

A monitoring device (not shown) for defining the angular position of the motor or of the rotor 21 is disposed on the other side of the motor on which the spindle is also extended. The device substantially comprises a rotary capacitor with rotatable plates which are mounted on the spindle and fixed plates which are mounted in insulated manner on the casing and extend into the spaces between the rotatable plates. The angular position of the motor or of the rotor 21 may be clearly determined by the variable capacitance if a voltage is applied between the rotatable and the fixed plates. The device for defining the angular position will not be described in detail in this context because the manufacture of such a device is obvious to the expert.

The operation of the regulating valve is based on the fact that the rotor 21 is rotated by the servomotor on the basis of control signals so that the rotor bore 22 is optionally moved over one of the casing ports 23 and 24. Respiratory gas which is pressurized is supplied through the inlet duct 1. At the beginning of the inhalation phase, the rotor 21 is rotated so that the rotor bore 22 is disposed above the casing port 23. The respiratory gas is therefore able to flow via the duct 3 to the inlet of the I/E valve. The flow and pressure of the respiratory gas may be steplessly variably controlled between zero and a maximum value by appropriate relative displacement between the rotor bore and the casing port 23, this varying the effective cross-section of the aperture.

At the beginning of the exhalation phase, the rotor 21 is rotated so that the rotor bore 22 is disposed above the casing port 24. The gas will then flow via the duct 4 to the second input of the I/E valve where it is used, as will be shown subsequently, to generate a negative pressure to assist exhalation. The flow and pressure of the gas which flows in the duct 4 are of course also steplessly variable in this position between zero and a maximum value.

It may be desirable for intervals to be interposed between the two phases of a respiratory cycle, no respiratory gas being supplied to the patient nor exhalation being assisted by the generation of a negative pressure during such intervals. In this case, the rotor 21 is rotated so that the rotor bore 22 is not in alignment with either of the two casing ports 23 and 24. The position between the two casing ports 23 and 24 is appropriately selected in this case in order to minimise the angle of rotation of the rotor. The construction of the valve 2 does not incorporate any seals with friction because these would substantially impair the operating speed. The very slight distance between the external surface of the rotor 21 and the internal wall of the casing 16 achieves sufficient sealing tightness, that is to say respiratory gas losses are insignificant. Moreover, the venting port 26 prevents

respiratory gas passing via the motor chamber into the monitoring device where it might result in defects.

As already mentioned, the plastics material block 25 containing the ducts 1, 3 and 4, is part of a housing in which the I/E valve is also disposed. Fig. 4 shows a section along the longitudinal axis of the I/E valve and passing through the regulating valve 2, the I/E valve 5 and the measuring head 9. The I/E valve has a first duct or main duct 28 and a second duct or auxiliary duct 29 extending perpendicularly to the first mentioned duct and merging therein approximately in the middle thereof. One end 30 of the main duct 28 serves as inlet and communicates directly with the duct 3 which extends from the regulating valve. The other end 31 of the main duct provides communication between the I/E valve and the inlet duct 6 of the measuring head. In its inner zone adjacent to the main duct, the auxiliary duct 29 has initially a cylindrical shape and flares into an expanding taper in an adjoining zone 32. The end 33 represents the outlet of the I/E valve and is provided for connection to the duct 7.

An elongated projectile-shaped member 34, described hereinafter as a flow-immersed member, is mounted in the main duct 28 coaxial thereto and disposed in the zone in which the auxiliary duct extends into the main duct. The cross-section of the flow-immersed member 34 has substantially three zones. In the first zone 35 nearest to the inlet 30, the member 34 has a tapered shape with a spherically rounded end surface, the middle zone being cylindrical and the adjoining third zone 37 once again having a tapered shape with a spherical rounded portion at the end. The position at which the cylindrical zone 36 merges into the third tapered zone 35 is disposed in a plane which extends through the axis of the auxiliary duct and is perpendicular to the axis of the main duct.

A bore 38 extends from the spherically rounded endface of the flow-immersed member 34 nearest to the measuring head, initially coaxially to the main duct 28 and then bent through 90° extends coaxially to the auxiliary duct 29 as far as the part of the outer or external surface of the flow-immersed member 34 which is disposed opposite to the position at which the auxiliary duct extends into the main duct.

A thin tube 39 is inserted into corresponding bores which are coaxial with the auxiliary duct through the casing wall disposed opposite to the merging position of the auxiliary duct and through the flow-immersed member. The tube 39 merges into the cylindrical zone of the auxiliary duct 29.

In the zone of the first conical part 35 nearest to the inlet 30, the flow-immersed member 34 is provided with a plurality of vanes or guide walls 40 disposed in a ray-pattern and set at a slight angle, for example 6°, relative to planes extending through the axis of the main duct and intersecting the aforementioned guide walls. The vanes 40 are provided for mounting and centering the flow-immersed member 34 while their inclined position imparts twist to the respiratory gas which passes through the pipe.

The space 41 between the flow-immersed member 34 and the duct 6 functions as a diffuser. The flared part 32 of the auxiliary duct 29 also functions as a diffuser.

As already mentioned, the control valve 2 supplies gas under a mean pressure of approximately 1.2 atm abs alternately through the ducts 3 and 4. During the inhalation phase, gas passes through the inlet 30 into the main duct and flows past the flow-immersed member 34. The reduction of the tube cross-section in the tapered zone 35 of the flow-immersed member 34 results in a substantial rise of flow velocity of the gas, that is to say its pressure energy is converted into kinetic energy. In the merging zone of the auxiliary duct 29 the respiratory gas will be at atmospheric pressure. Accordingly, there will be no pressure gradient between the main duct and the auxiliary duct and accordingly practically no respiratory gas will be discharged through the auxiliary duct. The cylindrical part 36 of the flow-immersed member is relatively long to enable laminar flow to form. Pressure energy is recovered from the kinetic energy of the gas in the diffuser zone 41 and a pressure of approximately 1.1 atm abs is obtained at the end 31 of the main duct.

The vanes 40 impart twist to the gas so that neat detachment is obtained at the end of the flow-immersed member while avoiding detachment from the wall of the main duct.

During the exhalation phase, the gas exhaled by the patient passes through the measuring head 9 (Figure 1) and the duct 6 into the zone of the diffuser 41 and through the bore 38 into the auxiliary duct 29. Since the free cross-section which remains between the cylindrical zone 36 of the flow-immersed member and the wall of the main duct is smaller than the cross-section of the bore 38, it follows that practically no exhalation gas passes by the flow-immersed member 34 to the inlet side of the main duct even without any additional measures being taken. The size of the bore 38 is made sufficiently large to permit exhalation without external aid. From the auxiliary duct 29, the exhaled gas

may be released to atmosphere or recycled.

During the exhalation phase, the regulating valve supplies the pressurised gas through the duct 4 to the tube 39 and through the said tube into the auxiliary duct 29. Owing to the smaller cross-section of the tube 39, the gas contained therein has a high velocity with which it also flows out into the auxiliary duct 29. This results in the action of a jet pump by means of which a suction effect may be produced on the patient side to assist exhalation. Extraction also assists in preventing exhalation gas passing by the flow-immersed body to the inlet side of the main duct. The kinetic energy of the gas is once again converted into pressure energy in the diffuser zone 32 of the auxiliary duct so that the gas is once again at atmospheric pressure at the outlet 33.

The I/E valve has two important advantages compared with known I/E valves. Since it does not contain any moving parts, it is impossible for functional defects to occur as the result of liquid droplets and solids entrained in the exhalation gas being deposited. In practice, the ducts cannot be blocked because they are always blown free during the succeeding inhalation phase. The second advantage is due to the fact that because of its simple construction, the I/E valve is constructed entirely of plastics and may be integrated with the duct block 25 of the regulating valve and the duct of the measuring head 9.

Fig. 4 shows the construction of the measuring head 9. This substantially comprises two parts of which the first part 42 contains the patient duct 6 through which respiratory gas flows during the inhalation and exhalation phases while the second part 43 contains the measuring device. The part 42 of the measuring head 9 which contains the patient duct 6 comprises a plastics material block and is part of the housing which also contains the I/E valve and the plastics material block 25. The second part 43 which contains the measuring device on the other hand is mechanically coupled to the regulating valve 2 or to the motor 18.

The housing comprising the plastics material block 25, the I/E valve 5 and the duct part 42 of the measuring head 9 represents an interchangeable assembly which is releasably attached to the fixed part of the respirator to which in the regulating valve 2, the motor and the monitoring device for defining the angular position of the motor and the part 43 containing the measuring device of the measuring head belong. The aforementioned interchangeable assembly is designed as a disposable part so that it may be replaced after being used once (that is to say for one patient) thus dispensing with the need for cleaning and sterilisation. However, the part is appropriately constructed from a plastics material which may be sterilised so that the usability of the respirator is ensured in an emergency when no new parts are available.

Ducts 44, one of which is shown in Figure 4, are provided to connect the patient duct 6 to the measuring device. One end of the aforementioned ducts 44 terminate at fixed measuring points of the patient duct 6 while their other ends are open to the outer boundary of the part 42. If the interchangeable assembly is assembled with the fixed parts, the aforementioned outer ends of the ducts 44 will be disposed opposite corresponding apertures 90 of the measuring device as shown in Figure 6 and the said outer ends are sealed with respect to the ambient air as described hereinbelow.

The principle of the measuring head will first be described by reference to Fig. 5. Fig. 5 is a plan view of the interchangeable part 42 and shows in diagrammatic form the measuring device of the measuring head 9.

The part 42 comprises a substantially L-shaped plastics material block of rectangular cross-section. The patient duct 6 extends in the interior of the said block as a duct which is substantially bent through 90°. The first part 45 of the duct 6 is connected to the I/E valve and has a circular cross-section followed by a transition zone 46 to merge into a rectangular cross-section. The wall 47 which is the curved outer wall of the duct initially extends paralel to the axis of part 45 and continues from a specific plane which is perpendicular to this axis in a bend 48 with a radius of curvature which is identical to the longer side of the rectangular cross-section. In a second part of the duct 6, whose axis extends perpendicularly to the previously mentioned axis, the outer wall 49 at first proceeds also parallel to the axis of this second part and then continues in a bend 50 which is accurately symmetrical to the bend 48. The wall on the inside of the bend or curve has two portions 51, 52 disposed parallel to portions 47, 49 of the outer wall while a part which is disposed opposite to the bends 48, 50 extends at an angle of 45° but without any curvature.

Four measuring points to which connecting ducts 44*a*, 44*b*, 44*c*, 44*d* extend are provided for measuring flow and pressure. As already mentioned, these ducts 44 connect the patient duct 6 and the measuring device. The two connecting ducts 44*a* and 44*b* merge into the patient duct at positions at which the portions 51, 52 of the inner wall are bent into the portion which extends at an angle of 45° to portions 51 and 52. The two other connecting ducts 44*c* and 44*d* terminate on the outer

wall of the duct 6 at a position at which the axes of the two parts of the duct 6 meet the curved wall parts 48 and 50.

The measuring device is provided with two ducts 53, 54 which on the one hand communicate with the connecting ducts 44*a*, 44*b* and on the other hand communicate with each other through a central cavity 55. Two further ducts 56 and 57 extend from the cavity 55 to the connecting ducts 44*c*, 44*d* which terminate in the outer wall of the patient duct 6. At the two ends of the two ducts 56, 57, that is to say near the cavity 55 and near their point of connection to the connecting ducts 44*c*, 44*d* the two ducts 56, 57 are provided with pneumatic orifices 58, 59 or 60, 61 respectively. Temperature-dependent resistors (referred to hereinafter briefly as thermistors) 62, 63 are mounted immediately adjacent to the orifice aperture and as far as possible concentrically thereto between the two connecting ducts and the two diaphragms 59, 61 which are disposed on the side of the connecting ducts 44*c*, 44*d*. The connecting leads of the thermistors 62, 63 are brought out laterally from the ducts and are connected to the electronic part of the measuring device which will be described subsequently.

The two ducts 56, 57 are also connected respectively to ducts 64 and 65 each of the said ducts in turn communicating via pneumatic resistors 66, 67 with a common duct 68. The duct 68 is connected to a source of pressurised gas (not shown) which is independent from the inhalation gas supply.

A further duct 69 extends from the central cavity 55 to atmosphere. The said duct contains pneumatic orifices 70, 71 disposed near the cavity 55 and near the duct end adjacent to atmosphere and two pneumatic resistors 72, 73 disposed between the aforementioned two orifices. A thermistor 74 is mounted between the pneumatic orifice 70 and the cavity 55 directly adjacent to the orifice aperture and a thermistor 75 is mounted between the pneumatic diaphragm 71 and atmosphere directly adjacent to the orifice aperture. The part of the duct 69 disposed between the two pneumatic resistors 72, 73 communicates via a duct 76 and a further pneumatic resistor 77 with the duct 68.

Finally, a duct 78 extends from the cavity 55 to atmosphere. A narrow duct 79 communicating with the duct 76, and a narrow passage 80, opposite to the duct 79, extends into the duct 78; a thermistor 82, extending to atmosphere and disposed downstream of an orifice 81, being disposed in the said narrow passage 80.

The connecting leads of the thermistors 74, 75 and 82 are also brought out laterally from the ducts and extend to the electronic part of the measuring device.

The method of operation of the measuring device is as follows: A stream of gas distributed over the individual branches of the device, is supplied from a source of pressurised gas via the duct 68. Part of the gas stream passes through the pneumatic resistors 66 and 67 into the ducts 64 and 65 and then into the ducts 56 and 57. Since the events which take place in the ducts 56 and 57 are identical, the description may be confined to one of the two ducts. namely 56. The stream of gas which arrives via the duct 64 is once again divided into two branches in the duct 56 one of the said branches flowing through the aperture of the orifice 58 into the cavity 55 while the other passes through the aperture of the diaphragm 59 past the thermistor 62 into the connecting duct 44*c* and from there into the patient duct 6. These two part streams are provided to measure the inhalation flow in the patient duct 6. If no gas flows in the patient duct 6, the same pressure will prevail at all measuring points 44*a*-44*d*. The two branch flows which flow through the apertures of the orifices 58 and 59 therefore meet the same pressure and remain unchanged with respect to time. During the inhalation phase, respiratory gas will however flow from the I/E valve through the patient duct and "strike" the curved part 48 of the outer wall. The pressure of this flow at the end of the connecting duct 44*c* is higher than in the zone of the inner wall of the duct, that is to say at the ends of the connecting ducts 44*a* and 44*b* and the pressure is therefore also higher than that in the cavity 55. Due to this pressure difference, the two part flows in the duct 56 vary in such a way that the stream passing through the aperture of the orifice 59 is diminished and the stream passing through the aperture of the orifice 58 is increased. This change of the branch stream which passes through the aperture of the orifice 59 is measured by means of the thermistor 62.

The impedance of the thermistor is adjusted to a constant value for measurement. The thermistor is uniformly cooled by a uniform gas stream which passes through the orifice 59 so that a state of equilibrium is obtained under steady-state conditions. Under non-steady conditions, that is to say when inhalation flow is present and the branch stream which passes through the aperture of the orifice 59 is altered, the heat dissipation caused by the gas is also altered and the power losses of the thermistor 62 are correspondingly altered. The electric current supplied to the thermistor 62 is measured and a change of power loss is thus detected.

It has been found that the change of the ratio of the two branch flows in the duct 56 and therefore the heat dissipation produced on the thermistor 62 by one of the two branch streams and the change of power loss of the thermistor or the signal indicating such a power loss change are unequivocally related to the change of the inhalation flow which prevails in the patient duct 6. An electric signal which indicates the flow may thus be obtained. The flow signal is supplied to the electronic circuit described subsequently for the purpose of regulation.

The exhalation flow is measured in the same way by means of the thermistor 63 disposed in the duct 57.

To this end it was assumed by way of simplification that no signal is obtained from the thermistor 63 during inhalation and no signal is obtained from the thermistor 62 during exhalation. This is not strictly correct but these undesired signals are substantially smaller than the signals employed for measurement and may be eliminated without difficulty in subsequent processing.

Part of the gas stream supplied through the duct 68 passes via the pneumatic resistor into the duct 76 and this part of the gas stream is utilised for measuring the gas pressure in duct 6. The stream enters the duct 69 and passes from there via the two pneumatic resistors 72 and 73 to the orifices 70 and 71. If atmospheric pressure prevails in the cavity 55, the gas streams which pass through the apertures of the orifices 70 and 71 and therefore the cooling effect of such gas streams on the thermistors 74 and 75 will be of identical magnitude. If there is a pressure difference between the pressure which prevails in the cavity 55 and the atmospheric pressure, there will also be a difference in the gas streams which cool the thermistors 74 and 75 thus enabling a differential signal to be obtained. It has been found that this differential signal is proportional to the required pressure difference.

Part of the gas stream which enters the duct 76 is conducted via the duct 79 into the duct 78 and is used for zero-setting of the pressure signal. Since the duct 78 provides for communication between the cavity 55 and the atmosphere, a flow in one or the other direction will be formed depending on the sign of the pressure difference between the pressure which prevails in the cavity 55 and is described as buccal pressure P_{buec} and the atmospheric pressure. If there is no pressure difference between the cavity 55 and the atmosphere and no gas therefore flows in the duct 78, this state corresponding to the desired zero, the gas which enters the passage through the duct 79 and through the aperture of the diaphragm 81 will reach the thermistor 82. Since the buccal pressure always alternates between a value below atmospheric pressure and a value above atmospheric pressure during respiration, it follows that the desired zero state always occurs briefly in the duct 78. The thermistor 82 delivers a signal because of being cooled during this brief period of time. This signal may be utilised for zero-setting the measured pressure signal for each respiratory cycle. This device for zero-setting dispenses with frequent manual zero-setting.

Zero-setting is required for the flow signals in the same way as for the pressure signals obtained from the system because the zero lines of both signals have a tendency to shift due to different external effects. However, the flow signals are zero-set by electronic means as will be shown subsequently.

While the device for measuring flow and pressure is shown diagrammatically in Fig. 5, the actual construction may be seen by reference to Fig. 4 and 6. All ducts as well as the cavity 55 are disposed in the interior of a metal block 83 comprising two parts which are placed flat upon each other. The said ducts are partially milled into the parting faces between the two parts of the metal block 83 and they partially comprise bores situated perpendicularly with respect to the aforementioned parting face. The section shown in Fig. 4 extends through part of the duct 68, the pneumatic resistor 66, the ducts 64 and 56, the pneumatic diaphragm 59, the thermistor 62 and the duct 44*c*. A description of the ducts leading to one measuring point as shown in this section is sufficient to explain the construction because regarding the other measuring points the duct system is basically equivalent to that just described.

The pneumatic resistors may comprise metal capillaries which are adjusted to the desired resistance value by squeezing at a specific position. They may be impressed in known manner into communicating bores provided between the individual ducts. The bore representing part of the duct 56 terminates in the endface of the metal block 83 which is disposed opposite to the interchangeable part of the measuring head 9, namely opposite to the connecting duct 44*c*. An annular groove or trough 84 is milled into the endface of the metal block concentrically to the end of the aforementioned bore. The internal diameter of the said groove is larger than the diameter of the bore so that an edge remains. A cap 59 is inverted over the aforementioned edge, the flat part of the said cap having a concentrically disposed aperture. This aperture represents the pneumatic orifice. A cylindrical sleeve 85, open at both ends and

comprising electrical-insulating material is also inserted into the aforementioned groove. A sleeve 86 of the same diameter is placed upon the sleeve 85. The thermistor 62 is clamped with its wires between the two sleeves 85 and 86. The connecting wires of the aforementioned thermistor extend to both sides and are soldered to a printed circuit board 87. The sleeve 86 extends through the circuit board 87 and bears upon a rubber plate 88. The rubber plate 88 is provided with an aperture 89 and seal lips 90 which surround the said aperture in annular form. The entire system is disposed in a casing which has an opening into which the rubber plate 88 is inserted, the said opening being disposed on the side of the interchangeable part 42. In the assembled state, the rubber plate is pressed with the seal lips 90 on the surface of the interchangeable part 42 to provide a seal relative to the ambient air.

To enable the metal block 83 to be thermostatically stabilised, its other surface is provided with a space disposed between the metal block 83 and the wall of the casing 91 and sealed relative to the ambient air, the said space communicating with the supply of gas which is employed for measurement. Since nitrogen is normally employed for measurement this provides protection against explosion in this chamber which contains the power transistor 92.

As already mentioned, the respirator is controlled by the electronic control device 13 (Figure 1) which influences the regulating valve 2 on the basis of measured values of flow and pressure and on the basis of in-fed desired values. In the present embodiment, flow regulation takes place during inhalation and pressure regulation during exhalation under normal conditions. This kind of control has been found particularly appropriate for the following reasons: a specific preset volume of respiratory gas is supplied to the patient during inhalation. At the beginning of exhalation, the final inhalation pressure is taken over as starting value for pressure regulation and the buccal pressure is brought to zero or atmospheric pressure in accordance with a certain curve. The curves of flow and pressure characteristics are prescribed to the system in the form of desired values; the corresponding measured values are used to correct the desired curves. Maintaining the selected desired volume per minute was defined as an over-riding condition. To this end, the volume is calculated by means of flow metering and is used to correct one or more of the other set-point values.

Another condition was that the buccal pressure and its increase with respect to time must not exceed maximum values during inhalation. The control device changes over to pressure regulation if one of these maximum values is reached during the flow regulation in the inhalation phase until the flow curve permits pressure or a pressure rise below the maximum values.

The construction and method of operation of the control device are illustrated in Fig. 7 which shows a block diagram of this device. Fig. 7 and the description hereinbelow does not deal with details of the circuit because the functions illustrated by blocks may usually be obtained in several different ways which are well known to the expert. It is the combination of the individual functions, more particularly the combination of the individual functions, more particularly the combination into a combination into a plurality of subordinate control circuits, which is essential.

Fig. 7 first shows the control system comprising the regulating valve 2, the I/E valve 5 and the measuring head 9. The first of the subordinate control circuits regulates the position of the servomotor 18 (Figures 2 and 3) of the regulating valve. To this end, the motor 18 is connected to a circuit 100 for position control, hereinafter briefly referred to as the position controller, the variable capacitor disposed on the motor spindle and associated with the monitoring device being connected into the aforementioned circuit 100 which thus constitutes a closed loop. The position controller 100 is connected to a circuit 101 for selecting between flow regulation and pressure regulation, hereinafter briefly referred to as the selector switch. The selector switch 101 has three inputs of which the first is derived from a timing generator 102 which produces a time programme for the respiratory cycle on the basis of in-fed values. The other two inputs of the selector switch 101 are connected at two subtracting junctions 103′, 104′ to the outputs of two wave-form generators 103, 104 and to the returns of the flow and pressure control circuit. The wave-form generators produce the flow and pressure wave-forms for the respiratory cycle on the basis of in-fed values and correction values obtained from within the control device, the inhalation and exhalation phase proceeding in accordance with the aforementioned curves. The subtracting junctions 103′, 104′ are formed substantially by subtraction circuits in which the measured signals are subtracted from the desired value signals to produce an error signal. The loop, comprising the selector switch 101, the position control 100, the control valve 2, the I/E valve 5, the flow measuring device 105 and the adding junction 103 represent the flow regulating circuit.

The adding junction 104′ is correspondingly connected to the output of the pres-

sure measuring device 105′ thus forming the pressure control circuit.

The wave-form generators 103, 104 are provided with a plurality of inputs of which one each is connected to an external operating control 106 for selecting the desired wave-form. The desired wave-form defines the flow or pressure characteristic which would be maintained without interference. The selected wave-form is modified in accordance with requirements if disturbances occur. A second input of each curve of wave-form generators 103 and 104 is connected to the timing generator 102 which supplies the time programme, i.e for the length of time of a respiratory cycle and the ratio of inhalation and exhalation phase.

A third input of the flow wave-form generator 103 is connected to an external operating control 107 for selecting the desired volume per minute. The amplitude of the flow curve depends on the choice of the desired volume per minute.

A third input of the pressure wave-form generator 104 is connected to a logic circuit 108 which processes disturbances such as spontaneous respiration, coughing, exceeding the maximum pressure or maximum pressure rise and is used to initiate specific changes in the respiration cycle routine. For example, a changeover to pressure control takes place during inhalation if the maximum pressure or the maximum pressure rise is exceeded. On the other hand, the respiratory cycle is interrupted in the event of coughing or spontaneous respiration and resumed at the end of such a disturbance. To this end, a second output of the logic circuit 108 is connected to the timing generator 102.

The wave-form generator 104 is provided with a fourth input which is connected to a measuring device 109 for measuring the final inhalation pressure. A signal representing the final inhalation pressure is fed to the pressure wave-form generator 104 because it is to be utilised as the starting value for the pressure wave-form which is to be followed during exhalation. Any pressure wave-form jump which would be most undesirable is thus avoided. The input of the measuring device 109 is connected to the output of the pressure measuring device 105′.

The timing generator 102 has three inputs of which one is connected to the logic system 108 as already mentioned. A further input is connected to an external operating control 110 for selecting or adjusting the respiratory frequency. The third input of the timing generator 102 is connected via a I/E time ratio correcting circuit 111 to an external operating control 112 for the selection or adjustment of the time ratio between inhalation and exhalation phases. This preset ratio is varied in accordance with requirements by means of the correcting circuit 111. To this end, the correcting circuit 111 is provided with a comparison circuit 113 to compare the actual gas volume per minute supplied to the patient with the preselected volume per minute. The measured volume is computed in an integration circuit 114 from the measured values of the exhalation flow. The exhalation flow was selected for computation because the effect of any possible leakage during inhalation is already eliminated when such flow is used. The computed volume may be supplied via an output 115 to an indicating or recording device.

The logic circuit 108 is connected to a comparison circuit 116 which compares the actual pressure measured in the patient duct and the maximum permissible pressure as well as the rate of rise of the actual pressure and a corresponding maximum value. The two maximum values for pressure and rate of pressure rise are fed-in by means of external operating controls 117 and 118. The comparison circuit 116 is connected to the pressure measuring device 105′ for detecting the measured values.

Means for limiting the buccal pressure and the rate of pressure rise of the buccal pressure were provided because an excessive pressure or excessively rapid pressure rise could cause pain to the patient. For example, the maximum values must be set as low as possible in the event of injuries to the ribs or similar injuries.

The pressure signal supplied by the pressure measuring device 105′ is supplied to an indicating or recording unit 119.

As already mentioned, automatic zero setting is provided for flow measurement and pressure measurement. Automatic zero setting is incorporated in the flow and pressure measuring devices 105, 105′. Automatic zero setting whose zero setting signal is pneumatically obtained has already been described for pressure measurement. Zero setting is performed fully electronically for flow measurement. To this end, the circuit shown in Fig. 8 is provided which is contained in the flow measuring device 105. Reference should be made to Fig. 9 for a description of the operation of this circuit, this illustration showing the signals at different circuit points designated *a-h*.

The signal *a*, obtained substantially from the thermistor 62, and representing the inhalation flow is obtained at the input 120. This signal is idealised to the extent that no pressure change was assumed at the measuring point during exhalation. In actual fact, an amplitude which is substantially smaller than that during the inhalation phase is also obtained during the exhalation phase. Since the signals are added or

subtracted as will be shown subsequently, it follows that the small amplitude may be neglected in the phase which is not measured.

The signal *b*, obtained substantially from the thermistor 63 and subject to the same statement which applies to the signal *a*, is obtained at the input 121 and represents the flow during exhalation. The two signals *a* and *b* have no definitive zero line. A separate connection extends from the two inputs 120 and 121 to a subtraction circuit 122 and a separate connection to the addition circuit 123. The signal *g* which substantially represents the flow signal but has no difinitive zero line is obtained at the output of the subtraction circuit 122. The signal *c* is obtained at the output of the addition circuit 123. The output of the said addition circuit 123 is connected to an audio element 124.

The input of the audio element is connected via a capacitance 125 to the base of a transistor 126 whose emitter is earthed. A resistor 127 is connected between the base of the transistor 126 and earth. In this circuit, the signal at the base of the transistor 126 is represented by wave-form *d* and the signal at its collector, being the output of the audio element, is represented by wave-form *e*. An operations amplifier with a similar function was provided in the present case to improve accuracy.

The aforementioned signal *e* is supplied to a monostable multi-vibrator 128 from whose output the signal *f* is obtained. The signal *f* operates a switch 129 which briefly short-circuits to earth the signal obtained from the subtraction circuit 122 with each change over from inhalation to exhalation phase and viceversa. The output of the switch 129 provides the signal *h* which in this way is provided with a clearly defined zero line. The output signal *h* is the signal which is fed back to the input of the control circuit 101 for the purpose of flow regulation and is fed to the integration circuit 114 for the purpose of volume computation.

The system of subordinate control circuit provides a very wide bandwidth which, although not fully utilised, improves the accuracy in the operating range.

WHAT WE CLAIM IS:—

1. A respirator with automatic control for flow and pressure of respiratory gas, which respirator comprises a flow and pressure measuring device disposed directly adjacent to a patient connection, the device being for measuring the flow and pressure of respiratory gas and for converting these parameters into electrical signals, a valve system disposed between the respiratory gas source and the flow and pressure measuring device for controlling the flow and pressure of respiratory gas during inhalation and exhalation, and an electric control device comprising the measuring device, the valve system and the electric control device being interconnected to constitute a control loop for comparing the electrical signals generated in use of the respirator with set values and for generating a correcting signal for influencing the valve system.

2. A respirator according to Claim 1, wherein the valve system incorporates a regulating valve for changing over between inhalation and exhalation and for varying the flow and pressure of the respiratory gas during the inhalation and exhalation phase and said valve system further contains an inhalation/exhalation valve with a path extending during the inhalation phase from the regulating valve to the patient connection and during the exhalation phase from the patient connection to an outlet.

3. A respirator according to Claim 2, wherein the regulating valve contains a rotary valve comprising a stationary hollow cylindrical member the wall of which contains two bores whose axes are disposed in a plane which is perpendicular to the cylindrical axis, and a rotatable hollow cylindrical member rotatably fitting into and disposed coaxially within the stationary part, the wall of the rotatable cylindrical member having a bore whose axis is also disposed in the said plane which is perpendicular to the cylindrical axis.

4. A respirator according to Claim 3, wherein the rotatable member is connected to the spindle of a servomotor.

5. A respirator according to Claim 4, wherein means for detecting the angular position of the rotatable member are connected to the spindle of the servo motor and therethrough to the rotatable member.

6. A respirator according to Claim 5, wherein the means for detecting the angular position contains a variable capacitor.

7. A respirator according to any one of the Claims 3 to 6, wherein the stationary member is disposed in the bore of a casing block which contains ducts which extend to the interior of the rotatable member and to the bores of the stationary member.

8. A respirator according to Claim 7, wherein the casing block is part of a housing which also contains the inhalation/exhalation valve and part of the flow and pressure measuring device.

9. A respirator according to Claim 8, wherein the casing is constructed of plastics material.

10. A respirator according to any one of Claims 2 to 9, wherein the inhalation/exhalation valve is provided with a first duct extending from the inlet connected to the regulating valve to a connection which is associated with the device for

measuring flow and pressure, a second duct merging perpendicularly into the first duct and connecting the first duct to the outlet of the inhalation/exhalation valve and a flow immersed member disposed concentrically relative to the first duct in the zone in which the latter merges into the second duct, the flow-immersed member being provided with a middle cylindrical portion, a tapered portion at each end of the cylindrical portion and a bore extending from the endface of the flow-immersed member nearest to the measuring device to part of the external surface of the flow-immersed member disposed opposite to the merging position of the second duct.

11. A respirator according to Claim 10, wherein a duct is provided disposed concentrically to the axis of the second duct and extending therein.

12. A respirator according to Claim 10 or 11, wherein guide walls disposed in a ray pattern are provided in the zone of the taper portion of the flow-immersed member nearest to the inlet and being twisted relative to intersecting planes extending through the axis of the first duct.

13. A respirator according to Claim 10, 11 or 12, wherein the middle cylindrical portion of the flow-immersed member extends from the merging zone towards the inlet.

14. A respirator according to any one of Claims 10 to 13, wherein the second duct is provided with a flared portion which functions as a diffuser.

15. A respirator according to any one of Claims 10 to 14, wherein the inhalation/exhalation valve is constructed of plastics material.

16. A respirator according to Claim 15, wherein the inhalation/exhalation valve forms part of a housing constructed of plastics material and containing part of the regulating valve as well as part of the device for measuring flow and pressure.

17. A respirator according to any one of the preceding claims, wherein the device for measuring flow and pressure contains a kinked duct whose wall, which is outermost with respect to the kink, contains two bends which are concave with respect to the duct and meet in the middle.

18. A respirator according to Claim 17, wherein the measuring device is provided with duct(s) associated with a source of pressure gas for introducing at least one gas stream into the kinked duct.

19. A respirator according to Claim 18, wherein two ducts associated with a source of pressure gas terminate in the zone of the bends and wherein thermistors for measuring flow are disposed in the aforementioned ducts.

20. A respirator according to Claim 19, wherein connecting ducts are provided between the ducts which contain the thermistors on the one hand and the source of pressure gas on the other and wherein pneumatic resistors are disposed in the connecting ducts.

21. A respirator according to Claim 19 or 20, wherein pneumatic orifices are provided upstream of the thermistors.

22. A respirator according to Claim 19, 20 or 21, wherein at least one further duct terminating in the zone of the inner wall is provided which communicates with the duct or ducts containing the thermistors.

23. A respirator according to any one of Claims 18 to 22, wherein a further duct is provided which connects the ducts terminating in the inner wall to atmosphere, the end of the further duct which is open to atmosphere and its other end being provided with thermistors and a connection disposed therebetween with the source of pressure gas for the purpose of pressure measurement.

24. A respirator according to Claim 23, wherein pneumatic resistors are provided between the duct which extends to the source of pressure gas and to the thermistors.

25. A respirator according to Claim 24, wherein a pneumatic resistor is provided in the connecting duct which extends to the source of pressure gas.

26. A respirator according to Claim 23, 24 or 25, wherein an additional duct which connects the duct or ducts which terminate in the inner wall to atmosphere is provided for zero setting of the measured pressure signal, the additional duct being provided with ducts which extend into the additional duct at positions disposed opposite each other, one of the ducts being connected with the connection, which extends to the source of pressure gas, while the other duct contains a thermistor.

27. A respirator according to any one of the preceding claims, wherein the control device contains a plurality of subordinate control circuits.

28. A respirator according to Claim 27, wherein one of the control circuits contains a circuit for controlling flow and pressure, the valve system and the device for measuring flow and pressure for the purpose of regulating the flow.

29. A respirator according to Claim 28, wherein the circuit for controlling flow and pressure is associated with a wave-form generator for producing a desired wave-form which defines the flow characteristic.

30. A respirator according to Claim 27, 28 or 29, wherein one of the control circuits contains a circuit for controlling flow and pressure, the valve system and the measuring device, for the purpose of regu-

lating pressure.

31. A respirator according to Claims 29 or 30, wherein the circuit for controlling flow and pressure is connected *via* a second input to a wave-form generator for generating a desired wave-form which defines the pressure characteristic.

32. A respirator according to Claim 31, wherein the wave-form generator is associated with operating controls for selecting the curve shape, the respiratory frequency, the desired volume per minute and ratio of exhalation to inhalation.

33. A respirator according to Claim 31, wherein the wave-form generator is associated with an operating control for selecting the curve shape, with a timing generator, with a logic circuit for taking account of disturbances and with a measuring device for measuring the final inhalation pressure.

34. A respirator according to any one of Claims 27 to 33, wherein the means for measuring flow contain a device for automatic zero-setting of the measured signal.

35. A respirator with automatic control for flow and pressure of respiratory gas, substantially as hereinbefore described with reference to, and as shown in, the accompanying drawings.

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COMPLETE SPECIFICATION

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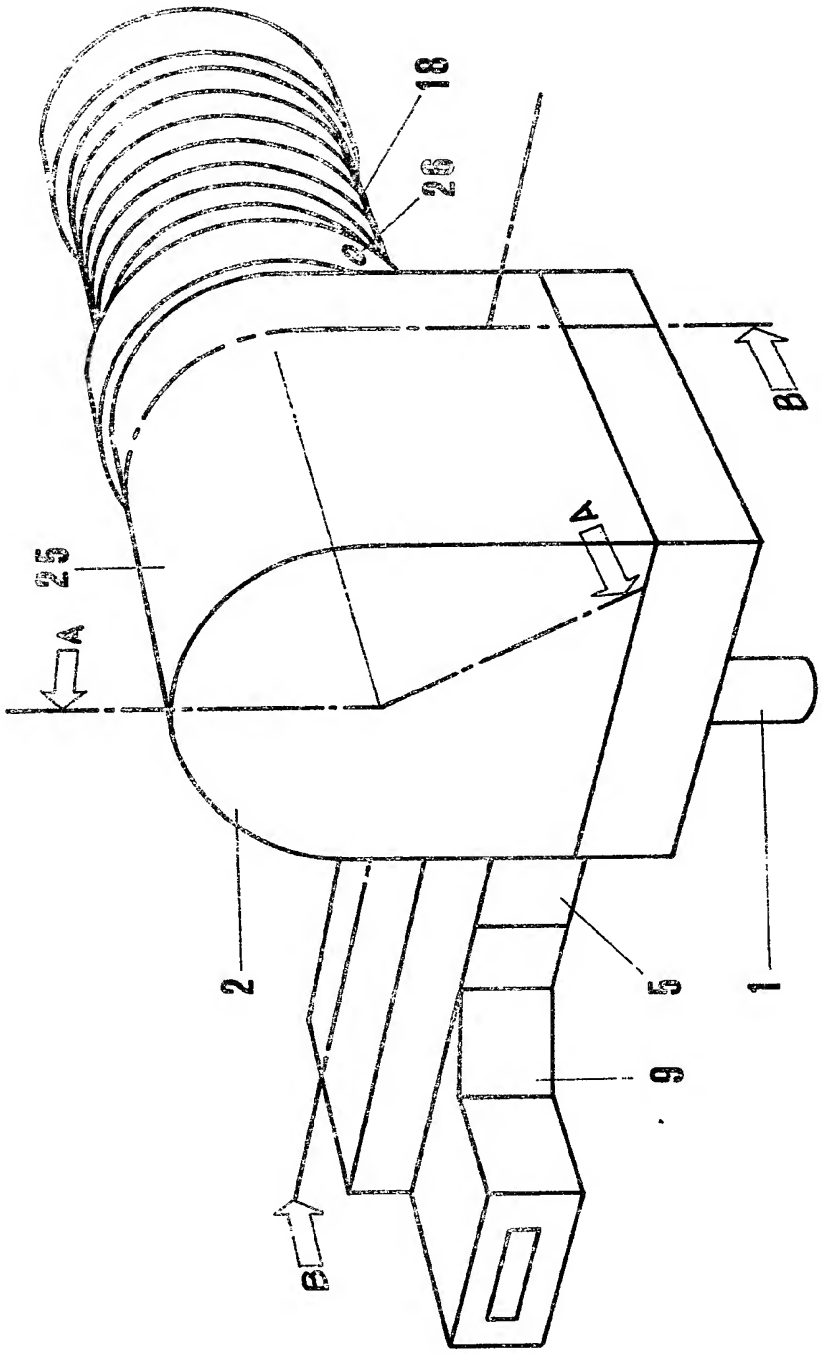

Fig. 2

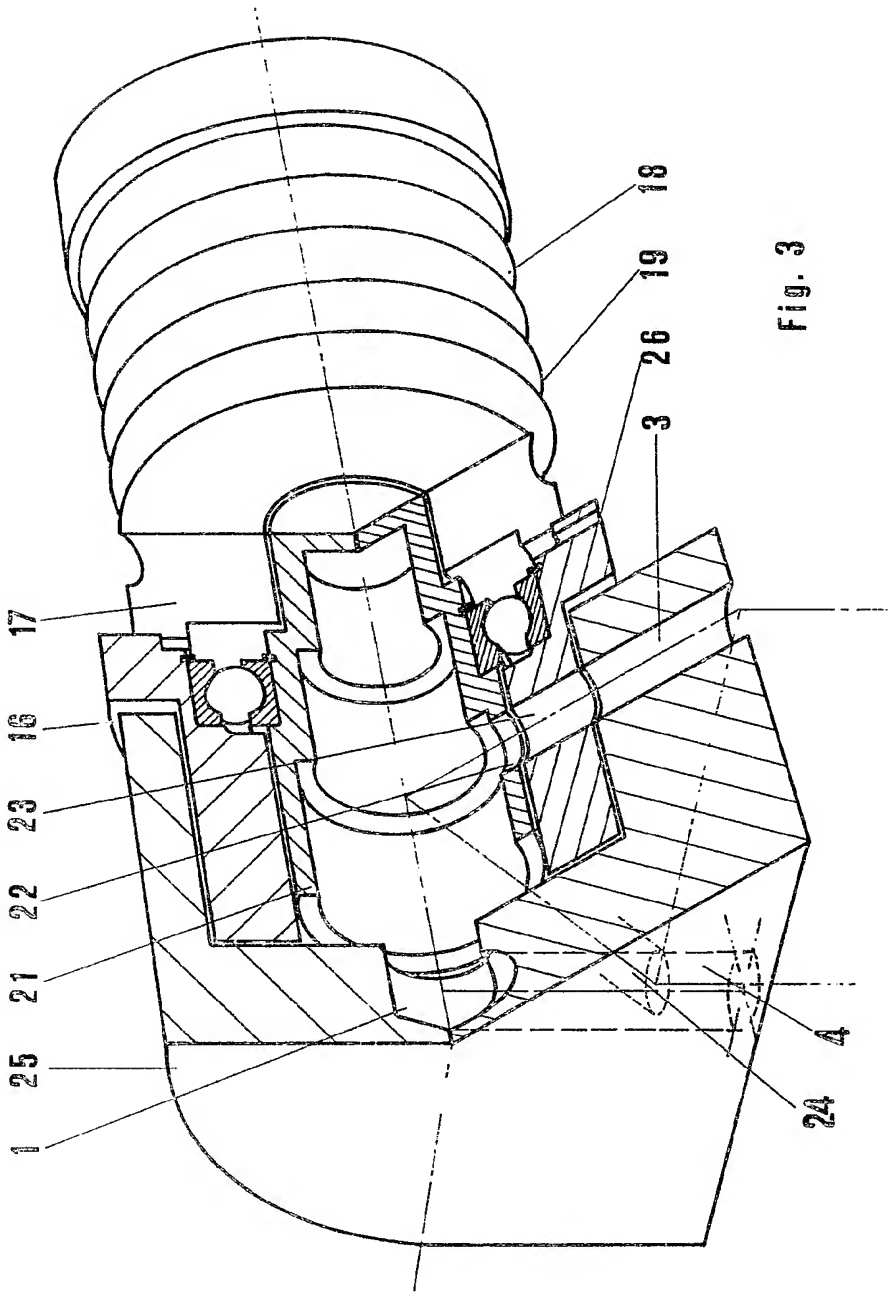

Fig. 3

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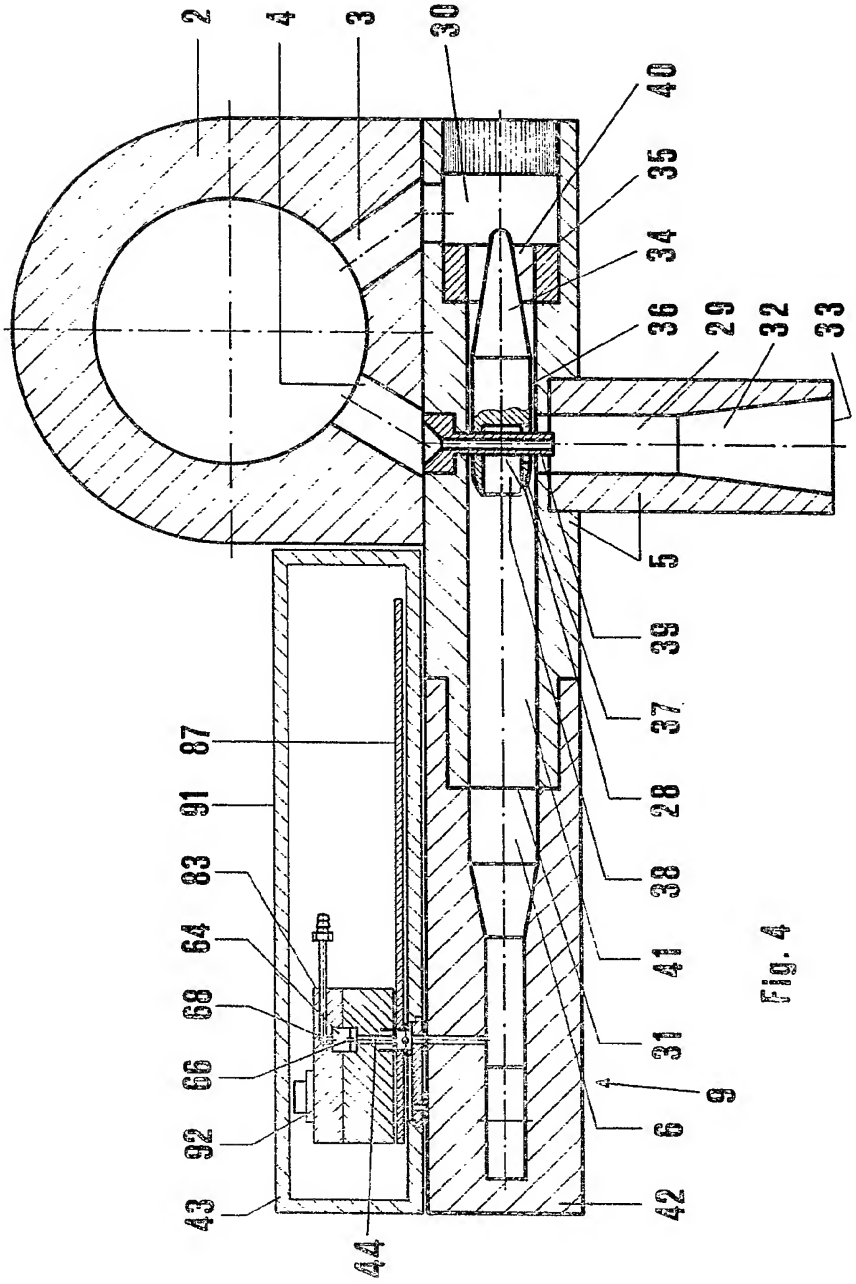

Fig. 4

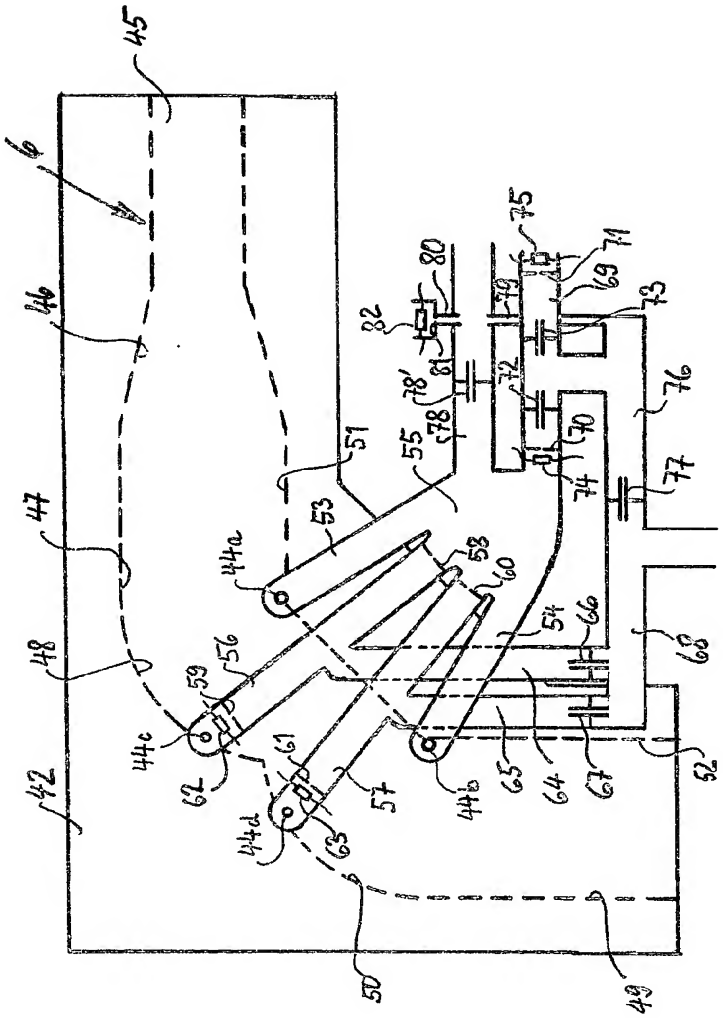
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Fig. 5

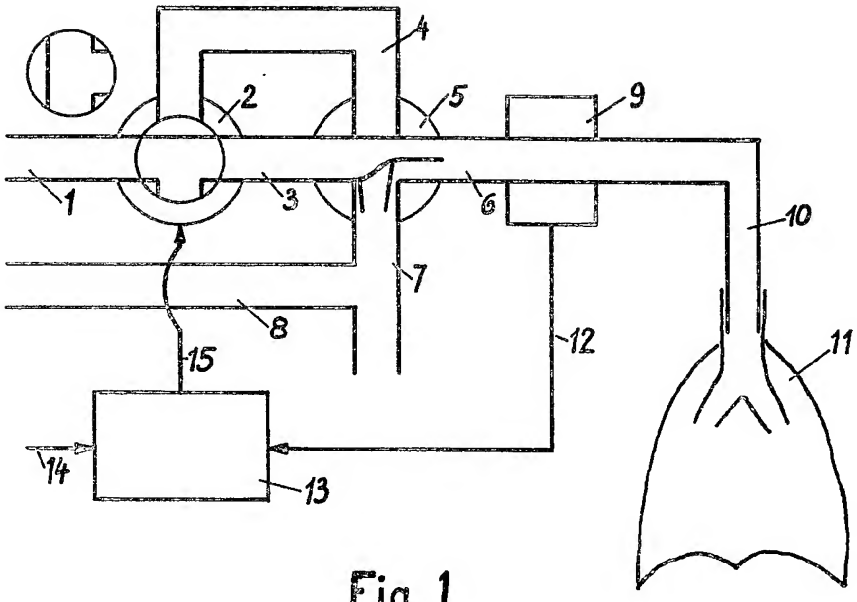

Fig. 1

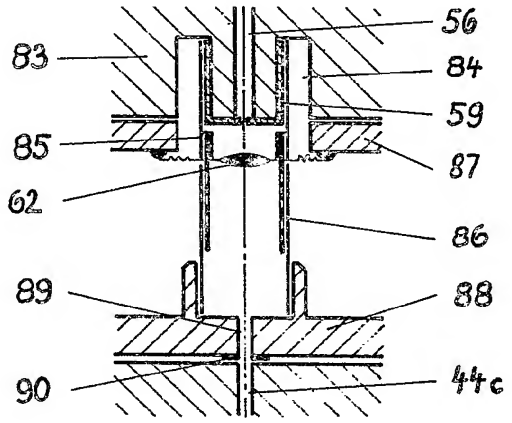

Fig. 6

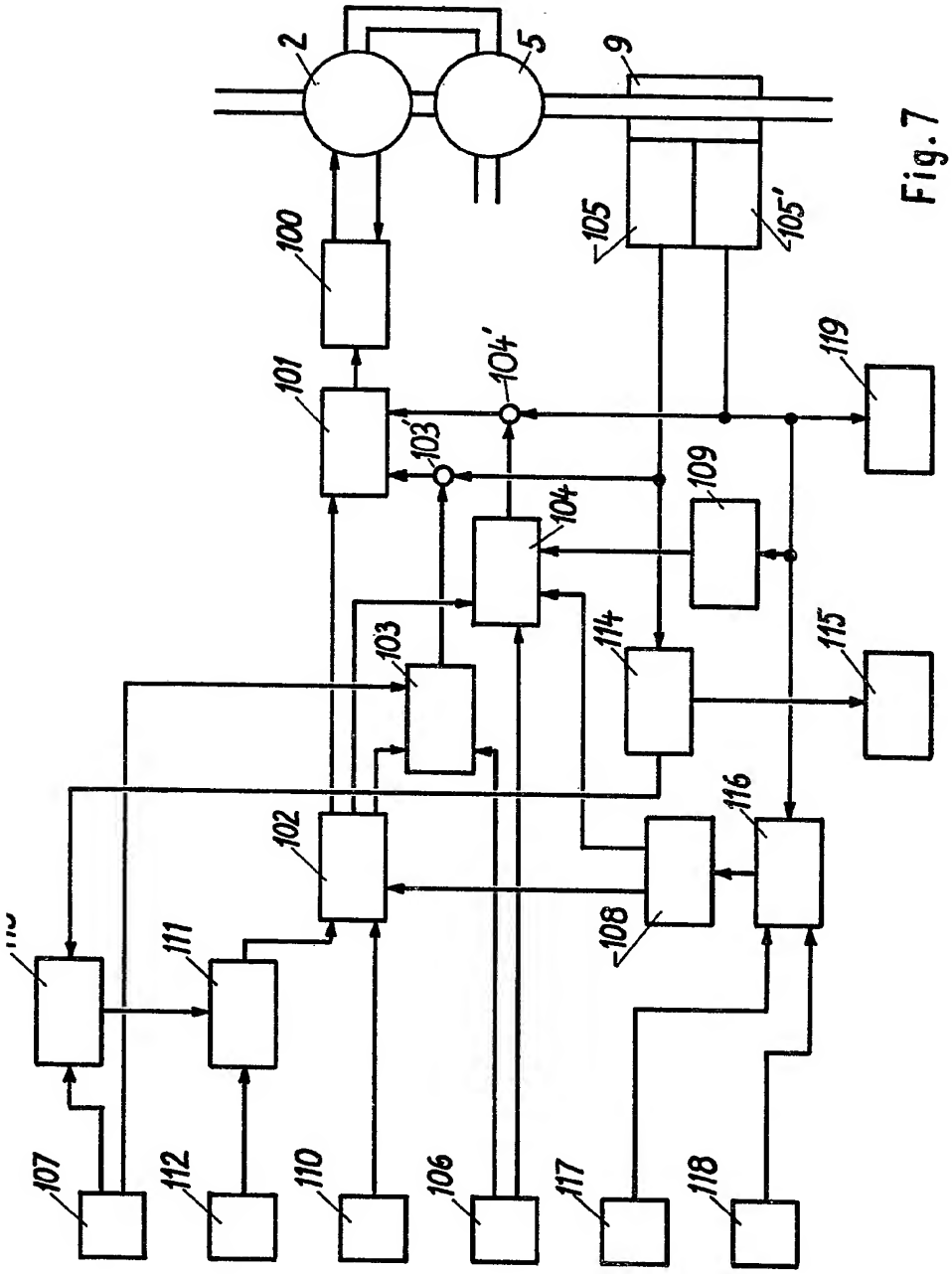
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Fig. 7

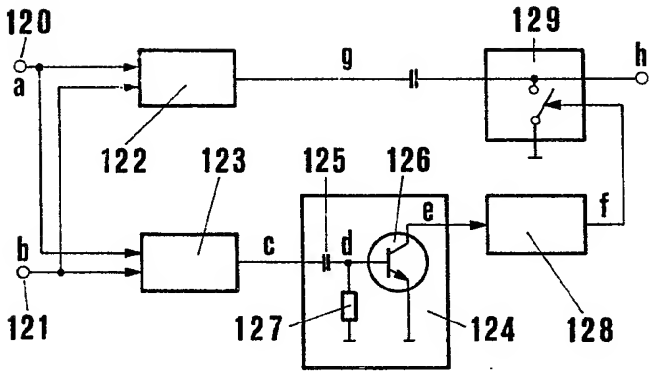

Fig. 8

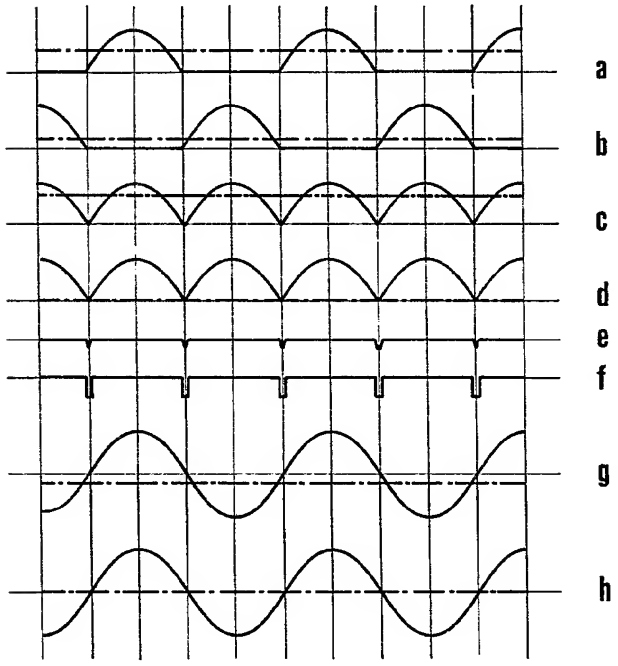

Fig. 9